BIO-POWER

INTERIM REPORT

October 15, 1966

Contract No.

Prepared by:

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I. STATEMENT OF OBJECTIVES

The object of this study is to survey past work in the area of body produced electrical power as reported in the literature, develop an understanding of the various methods and approaches which have been suggested as possible sources, and present this information in a useful form. This report is directed towards the general problem of providing electrical power to implanted electronic devices, such as artificial cardiac pacemakers, requiring less than 200 micro-watts for an indefinite time from a completely self contained internal system.

II. SUMMARY

A general literature review was first undertaken to both discover previous experimental work and to gather general impressions of thinking on this subject. The review indicated that there were few well documented conclusions available and that although a fresh start might be duplicative, it is needed for a systematic study. Conceivable sources were listed and each has been briefly considered from the physiological and instrumental viewpoints. In trying to evaluate potential usefulness of a source it was sometimes necessary to form conceptual designs of mechanical systems which could utilize that source. Comments on these designs are included in the discussion as contextual information because they illustrate those aspects of the systems which were considered and the design problems encountered, not because these designs are felt to be the proper solution. We concur with recent statements by investigators active in experimental work that the physiological Galvanic cell and mechanoelectric conversion are the two most promising systems, although much work is necessary before either is useful for practical purposes. Throughout this study we repeatedly encountered unanswered questions of the possible effects of biological adaptation on the total implanted system. These effects need not be entirely negative. Adaption in a direction tending to increase power output is a possibility deserving of serious attention.

III. GENERAL DISCUSSION

A. Directly Obtained Electricity

Since our goal is electricity produced from an energy source within the body, the simplest system instrumentally and the first which should be evaluated is the direct electrical tap. The systems under consideration in this section are all those that contain electrodes that are in physical contact with the biological environment and in which electrical potentials between electrodes can be measured. These potentials are due to, or at least intimately associated with, time and space variations in ionic concentrations and flow. The ultimate descriptions and explanations and definitions of these potentials are within the science of irreversible thermodynamics and the subject of considerable controversy. (19) For the purpose of this discussion we will generally consider these potentials operationally, that is, according to what you do to obtain a measurable potential and the characteristics of that potential.

1. Neuromuscular Potentials

The most studied biological electrochemical phenomena are the neuromuscular action potentials. At the cellular level micro-electrodes of several microns diameter inserted through neuron or muscle cell membrane indicate potentials of tens of millivolts between points within the cytoplasm and between cytoplasm and extracellular fluid. The time course of variations in these potentials is from a millisecond to hours. These "membrane potentials" in themselves are of no use for our purposes because of the fantastically low currents and short life of a cell damaged by perforation of its membrane. But

the gross extracellular phenomena associated with the summated effects of thousands of nearby active cell sections in nerve and muscle bundles deserve consideration. When a metal wire or disc is placed near a nerve, "compound action potentials" are observed between that electrode and another placed in the body, and when a muscle is used, the "electromyogram" is observed.

a. Nerve Potentials

The use of neural electricity as an energy source can be dismissed as not feasible on the grounds that drawing currents will cause stimulation of the excitable membranes and disruption of normal function. This conclusion results from observing that whatever electrical energy is produced is the result of neural activity (by assumption) and that that level of electrical current must of necessity be sufficient to cause stimulation since it was generated in order to produce (natural) stimulation. There is no way to avoid returning current to the electrode which supplied it when the voltages involved are less than the .4 volt minimum for germanium diodes, and while the two electrode current distributions will differ, it seems reasonable to expect that at least some units will be stimulated. The loss of even a few units in the PNS is undesirable. Also, since neural potentials are of the order of 100 millivolt, a source resistance of 12 ohm is needed to obtain 200 microwatts. This requirement is not compatible with the source. Consider for example, the input impedances (10^8 ohm) of the amplifiers typically required to measure nerve compound action potentials.

b. Muscle Potentials

With muscles the situation differs in several respects that make power pick up from this system at least conceivable. Namely, the volume of muscle is large compared with neural tissue and stimulation of a few units should not seriously disrupt normal function. The large muscle areas may allow many electrodes to be used simultaneously in order to provide a lower electrical source resistance. For example, if one pair of electrodes provides a source resistance of 500 ohm, 40 pair will provide 12 ohm. In order to utilize this low voltage ac source a miniature transformer can be used to step up the voltage before rectification. From a power standpoint it should be more efficient to use one transformer and rectifier with each pair of electrodes and sum the dc outputs. Since skeletal muscle EMG signals contain most of their power in frequencies above 100 cps, subminiature transformers ($\frac{1}{15}$ oz, $\frac{1}{40}$ in.^{3}) can be used. Answers to important biological questions were not found in the literature. No reports describing investigation of EMG signals as power sources were discovered, although use of EMG signals for control purposes have been frequently studied. Whether or not a low enough source resistance can be achieved and maintained and what type of electrode and implantation is best are questions which will probably have to be answered by experiment. Careful technique may prevent electrodes embedded in muscles performing large movements from causing irritation and pain but this will also require study. Judging from the experience of clinical workers with implanted cardiac pacemakers, small electrodes can be tolerated but dislodgement and lead breakage may be serious problems.

c. Cardiac Potentials

The strong rhythmic contractions of the myocardium attract immediate attention but the characteristics of its electrical potentials present particular difficulties. Unlike skeletal muscle which maintains contraction with a high frequency train of "spikes", cardiac muscle fibers depolarize and repolarize only once per heartbeat. Thus, the energy is contained mostly in low frequency components (1-20 cps). Since the contraction of all areas of the muscle is synchronized, large electrode areas can be used without the losses which would be associated with large electrodes on skeletal muscle. (Current from active non-synchronized skeletal muscle fibers could pass thru a large electrode to inactive tissue without passing to the second electrode of the pair.) But while this allows only one transformer to be used on the heart with one pair of large electrodes, transformers which are designed for very low frequencies are relatively large, heavy and inefficient. Regardless of the differences in waveform, the ECG on the heart is less than 100 mV (possibly only 10 mV) and thus a maximum source resistance of 12 ohm is again required for 200 uW power yield (assuming the 100 mV peak figure). Unless a radically improved electrode material is developed this resistance is not likely to be achieved. Effective source resistance figures for common electrode materials directly on muscle have not been found in the literature, and an experiment appears necessary to determine what resistance can be easily obtained and how this might change with time.

2. Direct Current Systems

These systems are all those that produce electron flow in one direction only between the two electrodes. This is unlike the neuromuscular potential electrode system in which a capacitor would be placed in the circuit to insure that no net charge flows from one electrode to the other. In the neuromuscular system the electrical energy of interest is " lternating current" in that charge flows from electrode A to electrode B and then, milliseconds later, returns to A. In the neuromuscular electrode system net current flow would provide no useful work and undesired reactions might accumulate products on or near the electrodes that result in "polarization" and/or electrode deterioration which interferes with the desired action. This type of electrode activity, however, can become the desired activity in direct current systems.

a. Physiological Electrode Cell

The Galvanic cell in its simplest and classical form consists only in two different metals (or other conductors) dipping into a common ionic solution. An electric potential, characteristic of the metals, temperature, ionic species and concentration, can be measured between the non-immersed portions of the two metal electrodes. Since the fluids existing within the body are ionic solutions inter-electrode potentials can be produced by inserting two dissimilar conductors anywhere. Because of the complexity of the body fluid composition, variation in composition at different points, induced effects from the presence of foreign material, presence of many membranes with unknown properties, active processes, etc., etc., the actual chemical reactions and inter-electrode potential can not be predicted, in fact such potentials are not strictly defined. (Nims, p. 3).

We can consider that there are two general types of reaction which occur with implanted electrodes. The first, which we will call type 1, is like that in the classical Galvanic cell described above in which one or both electrode materials enters into the reaction and becomes irreversibly altered or lost. The second type of reaction is possible when membranes are present and the chemical environment differs at the two electrodes. In this type 2 reaction, which in its simplest form is the classical concentration cell, irreversible change to the electrode surfaces need not occur and the electrodes need not be dissimilar. The rate of the chemical reactions may be much improved by dissimilar surfaces, however, for example by catalytic action, increased effective area, inducement of local environmental change, etc. This second type of activity can be considered as a fuel cell with the physiological system maintaining all the reactants and removing the end products. A special application of the type 2 system is when similar electrodes of "inert metals" or of "non-polarizable" liquid filled tubes are used in conjunction with very high impedance voltage measuring circuits which insure that the potential chemical reactions do not occur at the electrodes. This arrangement is used for investigating so called natural dc potential gradients within and on the surface of the body (4). Such measurements necessarily must draw virtually no power from the chemical energy sources responsible for the electrode potentials, for as soon as current is drawn reactions occur and the natural concentrations change. Thus any dc current producing electrode generator scheme useful for our purposes will have inter-electrode potentials which are more or less unnatural physiologically. That is, the potential difference between electrodes of any type drawing current will be different from that measured between "non-polarizable" electrodes drawing no current, and the local chemical environment surrounding the electrodes may be grossly

different. For these reasons the type 1 system in which electrode material change occurs and the type 2 system in which it need not are considered here as two special cases of the general physiological electrode cell system.

The general case of which the type 1 and 2 systems are examples, is when dissimilar metals, which produce a potential when dipped into a common ionic solution, are placed in a nonhomogeneous environment that produces a potential between similar electrodes. This is probably a fair description of the situation prevailing in the electrode material, electrode placement combinations with which Dr. John Konikoff and others produce the best results. The Konikoff work is a significant source of experimental data and has stimulated much of the recent interest in the physiological electrode cell power source. For these reasons a brief summary of the work reported in reference 12 is included here. The reader is referred to the original paper for details.

John Konikoff and Luther Reynolds were the principal workers at the General Electric Company's Space Sciences Laboratory under a contract with NASA in 1963 - 64 to investigate the use of what is referred to here as physiological electrode cell potentials as a biologically derived power source. Many combinations of electrode materials in several anatomical locations in several species of laboratory animals were tried. Their final choice of electrode materials was "high speed steel (75% Fe, 6% Cr, 18% W, .3% V, .7% C)" and a specially prepared "platinum platinum-black" combination. The final choice of location was as follows, "... the PPb electrode was located in the abdominal cavity dorsal to the peritoneal membrane and HSS situated subcutaneously but physically adjacent to the abdominal incision." The

longest continuous implant was 128 days, electrodes and sites were as above. The animal was a rabbit, and a constant resistive load of 10,000 ohm was applied between electrodes. After 15 days the output stabilized and thereafter remained at 24 microwatts and .5 volt. The highest power reported in short term studies was 308 microwatt. No new work from either Konikoff or Reynolds has been published since 1964. Telephone conversations with both men indicate that work is continuing, and that recent improvements in the platinum-black electrode material have increased the power output threefold for the same electrode area. Reynolds who is now at Hahnemann Medical College, Philadelphia reports that 200 microwatts has been obtained when each electrode is of 1/2 in^2 area. This electrode power generation scheme has the advantages, according to the originators, of simple surgical procedure and no harmful tissue reaction or loss of output at least for 4 months in the one long term rabbit experiment.

According to the data in the Konikoff report and especially the recent report of Strohl et. al. (29), when "biologically inert" metals such as platinum and type 316 stainless are implanted, power levels greater than 10 uW have not been obtained and the output drops significantly below this after a few days. Strohl's comments on the inevitable growth of a fibrous membrane around implanted electrodes suggests that the electrodes become isolated from the original, dissimilar ionic environments as this membrane grows. The better power outputs and longevity have been obtained in conjunction with an electrode which actively reacts with species present in the extracellular fluids. Even when covered with (hypothetical) cells tending to maintain identical ionic concentrations around the two electrodes, a reactive electrode can continue to provide current. In evaluating an electrode cell system containing reactive electrodes important considerations are toxicity of

products and deterioration of performance with time. As Strohl notes, Faraday's first law predicts the electrode weight loss due to ionic solution when the electrode reactions are known quantitatively. For example, .91 of iron will be needed to supply 100 uA for 1 year. But at least as important are the hard to predict effects such as loss of effective surface area "catalyst poisoning", uneven surface deterioration and long term local tissue reaction. In conclusion, it appears that there is a reasonable possibility that physiological electrode cells can provide 200 uW for extended periods, but careful long term studies and an understanding of the active phenomena, which, hopefully, will provide the basis for optimizing the electrode materials, are necessary. But the simple surgery in low risk areas which has been used, the mechanics - no moving parts, the non-dependence on any bodily motion, the inherent freedom from encapsulation problems, and the short term results already achieved, combine to make this a most promising system at this time.

b. Fuel Cell

Sophisticated direct current systems have been speculated on for producing relatively large quantities of electrical power for running proposed artificial hearts. These systems are usually referred to as fuel cells and usually are considered in reference to known chemical energy sources such as glucose or ATP. These systems are very appealing, largely because the proposed energy source is fairly well understood. Molecular energy yields, available concentrations and naturally occuring reactions can be stated. The development of physical systems to utilize these sources then appears to be a problem amenable to present technological capability since the available raw materials and necessary operations are known, at least

in broad outline. This is in contrast to the simpler Galvanic cell systems discussed earlier in which the present state of the art has been reached largely by trial and error without benefit of thorough understanding of the detailed processes involved. Approaching the problem from basic principles and proceeding in accordance with established theory will no doubt achieve practical success in time. The National Institutes of Health recently circulated a Request For Proposal to undertake feasibility studies of implanted biological fuel cells. When these initial studies are completed we will have a statement of the problem and outline of needed research. For the immediate future, however, the simpler a proposed system is, the greater appears its chance of success.

B. Thermoelectric Converter

Temperature gradients within the body theoretically can be exploited as a source of electrical energy. In recent years considerable research on thermoelectric compositions for use with nuclear reactor heat sources has produced materials with thermoelectric properties much improved over those of conventional thermocouples. For example, a conventional copper-constantin couple will produce 23 microvolt per fahrenheit degree temperature difference while a material of Bismuth-Antimony-Telluride composition produces 77 microvolt/F^{o} (11, 8). Simple calculations using this second figure indicate that with a 5 F^{o} temperature difference and 1 ohm resistance for every element 2500 elements connected in series will yield 200 uW at .5 volt. A Japanese group (32) has published a report of a 150 element thermoelectric generator for use on the external body surface. Their device used the Bi-Sb-Te material and the size of the thermoelectric array appears to be about 2.5 cm x 1cm x .5cm. The data presented in their report are not

clear and well organized and therefore the following calculations based on that report may not be completely correct. A maximum voltage of about 450 millivolt (open circuit?) is reported. A series array of 150 elements of a material producing 77 uV/F^{o} will produce 450 mV at a temperature difference of 39 F^{o}. Since some of their work was at 10^{o}C (53^{o}F) air temperature with evaporating alcohol on the cold junction, this temperature difference is possible. The only power output figure mentioned is 20 uW/cm^{2}. If this was obtained under conditions which produced a .45 volt open circuit voltage and if their device contacted a skin area of 2.5 cm^{2} then the indicated internal resistance of their device is 2000 ohms, or roughly 13 ohm per element. If this resistance figure is realistic for thermopiles composed of elements of 2mm x 1mm x 5mm size then the 2500 element array mentioned above would produce only 1/3 of the assumed 200 uW or only 15 uW. A total resistance of 13 ohm per element appears unnecessarily high, however, according to the following calculation. The resistivity of Bi-Sb-Te is only 7 x 10^{-4} ohm - cm (8). Hence an element of the above dimensions should have only 17 millohm internal resistance. Therefore the actual electrical resistance is almost entirely contributed by the contact between the thermoelement and the heat sink conductor and is largely a problem in technique. According to reference 26, contact resistivity in elements used in thermoelectric power generators may vary between 3 and 4500 microohm - cm^{2}. The higher figure indicates a contact resistance of .23 ohm for an area of .02 cm and, since there are two contacts per element, a total contact resistance of .5 ohm for elements the size of those in the Japanese device. This last calculation was the basis for our original assumption of 1 ohm per element.

The conclusion we reach is that a 2500 element array operating between a temperature difference of 5 F^{o}, with a surface area of 42 cm^{2},

at each heat sink and a depth of .5cm will produce 200 uW at .5 volt. By comparison, 25 cm^3 of medical grade mercury cells (8 Mallory RM CC-1W) (20) has a capacity of 8 AH which, neglecting age derating, will supply 200 uW for 5 years at 1.4 to 10 volt. Since the failure of any one of 5000 contact points in the series connected array will cause system failure, and since a 5 F^o temperature difference between two 42 cm^2 areas .5 cm apart does not naturally and reliably exist within the body, the thermocouple system is considered to be not competitive with conventional batteries for an implanted power source.

C. <u>Mechanoelectric Converter</u>

In this section possible mechanical energy sources will be considered together with mechanical coupling schemes. An arbitrary criterion of 1 milliwatt net mechanical work in the coupling system was chosen as a practical minimum power level for a final electrical output of 200 microwatt. Brief consideration of actual mechanical to electrical transducers, namely, piezoelectric crystals and permanent magnet generators, is included. We make the provisional assumption in this section that if a mechanical system can be implanted which will perform 1 mW work, for example in winding a spring, for over a year, then a transducer can be designed to utilize this energy. Other than work based on electrode cell potentials all known experimental implant power generation has been with piezoelectric crystals.

1. Sources

The obvious mechanical sources are:

- Voluntary muscle, joint and limb movements

- Peristalsis (dismissable on grounds of insufficient power)
- Respiratory system - rib cage motion, diaphragm muscle, thoracic and abdominal "pressure variation"
- Cardiovascular system - heart motion, aorta and large artery pulse expansion, blood flow
- Gross body acceleration (self winding watch principle, "random motion power")

Movements associated with voluntary activity in some cases offer large quantities of mechanical power. The intermittant character of this activity means, however, that an energy storage system must be included in the design to supply power during periods of inactivity. Rechargeable batteries are the obvious storage device, especially since they are designed for and require relatively high current, short duty cycle charging. These batteries require 50 to 100% more charge current than they return, however. Therefore, any intermittant generator will have to supply 300-400 microwatt average electrical charging power if the battery undergoes 200 microwatt constant drain. It may not be unreasonable to depend upon or require some particular voluntary movement being performed at some minimum rate for many months, but unless a particular application requires power only during a certain type of activity, it seems more straightforward to couple a motion generator to a continuous activity, such as respiration and blood flow, in which the rate and other operational norms and limits are predictable and unavoidable.

Respiration

The first continuous motion source which we will consider is respiration. Since the object of respiratory mechanical motion is to pump air, a fluid flow system operating on the pressure volume changes found in the thoracic and abdominal cavities during the respiratory cycle is an obvious possibility. During conditions of quiet rest the variation in pressure within the human adult thorax is approximately 3mm Hg (4 cm H_2O) or .04 Nt/cm^2. At a breath rate of 30/min, work of 2 millijoule per breath must be done for an average mechanical power of 1 milliwatt. If we approximate the phase lag to be expected between pressure and volume by assuming no phase lag but with only one half the pressure variation (i.e. .02 Nt/cm^2), then the volume of fluid (silicon oil, gas, isotonic saline, etc.) which must be pumped each breath according to the relation $PV = 2 \times 10^{-3}$ joule is 10 cm^3. Since the volume calculated in this manner is inversely proportional to breath rate and intrathoracic pressure, the volume required in most experimental animals will be less.

An elementary non-differential system responsive to respiratory pressure variations of 3mm Hg would probably be disabled by normal atmospheric pressure variations of one or two inches of mercury. Insensitivity to ambient "dc" pressure is inherent in a differential system, however, and because the intra-abdominal respiratory pressure variation is out of phase with the intrathoracic, two bellows, one in each cavity, connected by a tube would comprise such a system. It appears that this system can provide the necessary mechanical energy without obvious size and weight objections. A simple implantation procedure with a subcutaneous tube tunnel is conceivable, although all surgical questions as well as those on materials, size, shape and irritation require extensive

design and experimentation. In summary, a respiratory fluid pumping system is recommended as deserving of further attention.

Direct mechanical coupling to respiratory motion remains as another possibility. The change in dimension of the rib cage and diaphragm are attractive. The method of coupling might be something working on the principle found in retracting tape measures. A cable is wound on a drum and a spring tends to keep the cable wound up. If the drum package is firmly attached in some convenient location and the cable held against the under side of the diaphragm or in a subcutaneous tunnel around the chest with the far end of the cable attached, then the drum would rotate back and forth during each breath. A ratchet drive to wind a second spring would allow for any "zero position" of the cable extension with the second spring driving the actual transducer. Perhaps placing the cable inside a silicon rubber tube filled with silicon grease, the tube being of the bellows type to allow it to lengthen easily, would improve the sealing and tissue irritation situation. While quantitative data on the diaphragm has not been sought, it certainly appears that sufficient power is available from diaphragm motion and also from chest expansion. The main problems are expected to be in materials, packaging and surgical technique. Apart from material fatigue and sealing, tissue erosion and cell destruction from too great applied pressures must be avoided, for even living tissue applying pressure unnaturally (e. g. an aneurysm) can erode its way through other tissue. The experience with bone plates, wires and other prostheses which have been used for many years shows that direct mechanical attachments to internal structures can be accomplished, however.

Cardiovascular System

The other continuous mechanical source is the cardiovascular system.

The work of Doctors Parsonnet and Kennedy demonstrates, at least for short periods, that the expansion of the great arteries and the movement of the heart can be tapped for mechanical power. Specific comments on these sources are included in the discussions of their experiments. In general, however, a significant design problem is concerned with accomodating long and short term variations in the properties of the system being coupled to. With the arteries some of these variables are changes, whether natural or induced, in the artery cross section, arterial wall elasticity, average blood pressure, systolic-diastolic differential pressure, postural configuration and relative direction of gravity.

Random Motion

Gross body acceleration operating on a mechanical system similar in principle to the self winding watch refers to voluntary motion, especially walking, and the criticism of non-continuous sources applies. The only detailed consideration of this system is found in Dr. Long's article (14). The advantage of this system is that all the operating parts can be enclosed in a hermetically sealed rigid box. Except for the weight involved this box could be attached to the diaphragm or heart to take advantage of the continuous motion in these locations. But in order to demonstrate that the weight is prohibitive, consider a mass of M kilograms being forced to move back and forth over a distance of 1cm according to the sine law at a frequency of 1 per second. The maximum velocity achieved will be $\pi 10^{-2}$ meters/second. The kinetic energy at this point is $\frac{1}{2} MV^2 = \left(\frac{\pi^2 M}{2} 10^{-4}\right)$ joule. If we could somehow utilize all this energy each cycle, the mass for 1 millijoule is approximately 2kg. While the force necessary to accelerate this mass is only .2 Nt, which

could probably be provided by the diaphragm, the force necessary to support this mass against gravity is 20 Nt, which the diaphragm could not support. The artificial heart discussants are seriously considering weights of this magnitude for long term implantation; thus we cannot a priori dismiss a random motion system as unworkable, but it does not appear to be competitive with conventional mercury batteries in power per pound.

2. Transducers

a. Piezoelectric Crystals

The mechanical energy to electrical energy transducer most often considered for use in biological power applications is the piezoelectric crystal. Manufactured crystals of lead zirconate, lead titanate (PZT) composition have far superior properties for power transduction than do natural crystals such as quartz and rochelle salt. These manufactured crystals are produced in a form known as "ceramic bimorphs". Quantitative data on the relevant characteristics of these peizoelectric ceramics have been developed as part of this study from information available from the Clevite Corporation (9, 10, 22). Data of this sort are necessary for evaluating the practicability and design requirements of this method of power generation.

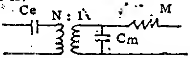
Efficiency

One of the most appealing characteristics of these transducers is an attainable conversion efficiency greater than 50%. This efficiency refers to the ratio of net mechanical energy supplied to the crystal to electrical energy supplied by the crystal under optimum conditions of mounting and matching. The simplest system for driving a crystal is to have the mechanical source directly coupled, that is, when the source (e. g. expanding aorta) moves, the crystal is deformed proportionately.

With this type of mechanical coupling a differently defined "efficiency" is significant. Net mechanical work means total work done on the crystal to deform it minus the total work done by the crystal as it relaxes to its unstressed state, $\left[\int_0^\infty F(t)v(t)\,dt\right]$. Since with direct coupling the source has to be capable of supplying enough work to deform the crystal, and since as far as the transducer is concerned any work it does on the source (during the relaxation phase) is lost forever, the ratio of total work done on the crystal,

$$\left[\frac{1}{2}\int_0^\infty \left[\left|F(t)v(t)\right| + F(t)v(t)\right]dt\right],$$

to total electrical work done by the crystal is a meaningful figure. The usual model for these crystals is:

Electrical  Mechanical

The Clevite literature contains tables for calculating: Ce, an electrical capacitance; N, the transducer ratio in volts per Newton; Cm, the mechanical compliance; and M, the mass, in terms of: the dimensions, L, W, T; specific crystal type, PZT-5B, PZT-5H, etc.; type of connection within the bimorph, series or parallel; method of mounting and drive, e.g. cantilever mounting with driving force at the free end. With the above model the internal stored electrical energy under short circuit conditions is $We = \frac{1}{2}\,CeV^2$ where $V = NF$ with F the force in Newtons. If an external capacitance C is connected and the crystal deformed by a force F, external work will be done in charging this capacitor. It can be shown that for the greatest external work, the external capacitor must be equal in value to Ce in which case $\frac{1}{4}$ We joules are supplied. Since in actual use a bridge rectifier would be used which allows an equal quantity of electrical work to be done as the crystal relaxes, $\frac{1}{2}$ We is the theoretical maximum electrical energy available. The mechanical work done on the crystal in deformation is approximately $\frac{1}{2}$ FD,

where d, the deformation, is CmF, thus $Wm = \frac{1}{2}\ CmF^2$. The ratio

is: $\frac{\frac{1}{2}\ We}{Wm} = \frac{CeN^2F^2}{2\ CmF^2}$ which for a PZT - 5B parallel bimorph

cantilever is $\frac{\left(2 \times 10^{-9} \cdot \frac{LW}{T}\right)\left(.3\ \frac{L}{WT}\right)^2}{(2)\left(2.8 \times 10^{-9}\ \ \frac{L^3}{WT^3}\right)}$ which reduces to

.032 or 3.2%. Of great significance is that all dimensions and the magnitude of force drop out! Also, a series connected cantilever bimorph, and an end - supported - center - driven mounting of either series or parallel connection can be shown to have the same efficiency, and the PZT - 5H material differs only slightly. Thus, while choosing dimensions, mounting, etc. will certainly effect the quantity of electrical energy produced in a direct-coupled system, this energy can never be greater than 3.2% of the mechanical work supplied (as defined above). Illustrations of the significance of these figures will be found in the discussion of previous experimental work.

Output

It was shown above that the maximum electrical energy which can be produced in one deformation cycle is $\frac{1}{4}\ CeN^2F^2$. The value of F, of course, can not exceed the force necessary to fracture the crystal. For a cantilever beam of length L and thickness T, the strain is

$$S = \frac{3\,T}{2\,L^2}\ D$$

where D is the distance the free end is displaced. Maximum strain before fracture is an intrinsic material parameter and one value holds (approximately) for any configuration. Mr. Carmen Germano of Clevite

has recommended 5×10^{-4} (50% of the fracturing strain)as the maximum strain to apply in a real system. We can assume a maximum deflection, then, of $D_{max} = \frac{2 L^2}{3T} \cdot 5 \times 10^{-4}$ for a cantilever mounted crystal and a maximum force of $F_{max} = \frac{D_{max}}{C_m}$ (D in inches implies Cm in in/Nt). Electrical output per deflection is, therefore, for a PZT - 5B parallel bimorph cantilever:

$$\frac{C_e N^2 D_{max}^2}{4\, C_m^2} = \frac{\left(2 \times 10^{-9}\ \frac{LW}{T}\right)\left(.3\ \frac{L}{WT}\right)^2 \left(\frac{1}{3}\ 10^{-3}\ \cdot\ \frac{L^2}{T}\right)^2}{(4)\left(1.1 \times 10^{-7}\ \frac{L^3}{WT^3}\right)^2}$$

$$= 420\ LWT \text{ microjoule,}$$

with L, W, T in inches. Since the bimorphs are only available in thicknesses close to 20 mil, the maximum output per deflection is 8.4 microjoule per square inch of area. Again, it can be shown that this figure is obtained for series as well as parallel connection, for center drive as well as cantilever.

The unit area output per deflection can be increased four times by allowing a "deflection" to be from - Dmax to + Dmax and back again, that is, by including a spring so that with no "external force" applied the crystal is deflected in the opposite direction from that which the external force produces. This variation in mounting does not effect efficiency since four times as much mechanical work is required. At the maximum strain of 5×10^{-4} with a spring loaded crystal, the absolute minimum surface area necessary to produce 200 uW is $\frac{6}{N}$ square inches, where N is the number of deflections per second, and, according to our previous definition of 3.2% efficiency, at least 6250 uW mechanical power to deform the crystal(s) is required. In order to extend the lifetime of the crystal by reducing fatigue it will probably prove

necessary to use smaller deflections than were assumed in the output calculations above. The output varies as the square of the strain however, so that at a maximum strain of 25% of the fracturing point $\frac{24}{N}$ square inches of area are required.

If added mechanical complexity is introduced (e. g. springs, ratchets, escapements, etc.) all of the work during the active phase (e. g. systole) can be retained in the crystal transducer package. While extra moving parts will cause losses themselves, some improvement in efficiency can be expected and the mechanisms can allow other important features to be incorporated, such as a improved crystal mounting and drive with a fluid-free environment. The diagram below will illustrate the type of system we had in mind.

Box	—	Bellows	—	Ratchet & Spring	—	Escapement	—	Striker	—	Crystals

A box of suitable material is fitted with a flexible metal "window" of the corrugated metal bellows type seen in wall barometers, which allows both mechanical motion transmission and the possibility of maintaining a fluid barrier. The bellows' movement winds a spring through a ratchet, which prevents any loss of energy back through the bellows. The spring drives a wheel with fingers that deflect the end of a cantilever mounted crystal. The alignment is such that when the crystal is deformed a preselected distance, the finger slides off and the crystal is set in oscillation at its natural frequency. All of the work done in the original deformation thus must be dissipated within the crystal, and electrical energy is available during each cycle of oscillation. The escapement prevents the next finger from engaging the crystal until sufficient time has passed for the oscillation to damp out. Since several

finger-pushes per second are possible, a relatively small area of crystal can be used. This arrangement also provides the desirable feature of allowing input energy to accumulate in the spring until sufficient force is developed to drive a finger over the crystal in the event that the input falls below normal. If excess input energy is available the crystal is driven more often, but, since its maximum deflection is always the same, it can not be broken. E. Van Haaften of Bulova Watch has designed a system similar in some respects to the above. (18)

b. Permanent Magnet Generators

A transducer not generally considered in discussions of implanted power is the permanent magnet generator. Because of the intrinsic weight of magnet and core materials, and because the usual mechanical input is rotational, this type of generator has little immediate appeal. However, if as has been suggested above, crystals require sealing in a total enclosure and drive through spring and gear mechanisms for optimum results then rotational input is not a relative disadvantage. If an implanted system must function for many years a p.m. generator should be satisfactory, while the fatigue lifetime of a crystal is not well understood. That weight is not an impossible obstacle is demonstrated by a generator manufactured by Rotating Components, Inc. and advertised in the 1966-67 Electrical Engineers Master catalog. This unit is 1.31" long by .95" diameter, weighs 2.5 oz. and, we calculate, can produce 100 milliwatt at 60 revolutions per second. A device especially designed for 200 uW output and low rotational speeds, therefore, should not be objectionably large, heavy or inefficient. We recommend that a p.m. generator not be dismissed until further data on the practical requirements and limitations of crystals become available. If the physiological and instrumentation problems of

obtaining mechanical work in the 10 milliwatt range can be solved, a p. m. generator may well be the better suited transducer.

3. Previous Work

Dr. John H. Kennedy

Dr. John H. Kennedy and Carl C. Enger at the Cleveland Metropolitan General Hospital have published several reports of their work with a self-powered cardiac pacemaker. (6, 7) The devices constructed and implanted by this group consist partially in a piezo-electric ceramic crystal mounted beneath a flexible plastic cover. The package containing the crystal generator as well as the rectifier and pacemaker electronics is sutured to the rib cage in a position where the beating heart applies pressure through the cover to the crystal. This system has, for short periods, provided effective pacing via stimulating electrodes in several experimental trials with dogs. Since our interest is in the power generating aspects, we will concentrate our attention on the crystal.

In the most recent paper (7) the crystal dimensions reported are 3.75 x 1.87 x .05 cm and drive parameters are described as, "... the mechanical energy needed to operate the self-powered pacemaker is 200 newtons or 21.4Cm." Let us assume that what was meant is a mechanical force of 2.0 newton. Using design equations supplied by Clevite, the manufacturer of the ceramic crystals, it can be shown that for an end supported, center driven crystal of the quoted dimensions, the force necessary to produce a strain of 5×10^{-4} (the recommended maximum) is 2.3 Nt. According to our calculations a PZT-5B crystal of the quoted dimensions deformed by a 2.3 Nt force will produce no more than 9.1 microjoule, and only 7 microjoule at 2.0 Nt. With a spring loaded crystal (and 4.0 Nt.) 28 microjoule is possible. Kennedy mentions the figure of

23 microjoule, but his reference is technically ambiguous. In order to provide a low ripple electrical source, it appears a larger capacitor than the optimum for maximum efficiency was used, however. An output voltage of 1.75 volts is mentioned. The maximum energy which can be supplied into a 1.75 volt source by a PZT-5B crystal with the above dimensions and 2.0 Nt driving force during a deformation - relaxation cycle depends on whether a series or parallel type of crystal bimorph was used. The best choice is parallel which can provide 5.4 microjoule. With a heart rate of 120/min the maximum power that could be produced is therefore 11 u W.

It is worth noting that the mechanical power necessary to deform a directly coupled crystal that provides 11 u W of electrical power is at least 340 u W. Therefore, if the transducer package can be designed to provide 20% efficiency rather than 3.2%, 68 u W could be obtained with no change in the mechanical power input. And if the displacement is increased from .2mm to .6mm, enough mechanical power is available to produce 200 u W at 20% efficiency. Even if only 10% efficiency is attainable, a deflection of 1.2mm to increase the power input does not appear unreasonable. Dr. Kennedy reports that after a one year implantation no damage to the adjacent myocardium was found. Electrical output apparently has not been maintained beyond a few days because of leakage of fluid into the package. Improvements in the crystal mounting and drive, in materials and packaging, and in surgical technique should be undertaken.

Dr. Victor Parsonnet

Dr. Victor Parsonnet and his co-workers at the Newark, New Jersey Beth Israel Hospital have published several papers describing their experiments with ceramic bimorphs mounted on the aorta. (e.g. 18, 21)

Their latest device contains two PZT-5 crystal slabs as the arms of a spring clothespin-like device which clamps onto the aorta. Each slab is $1\frac{1}{4}$ by $1\frac{1}{2}$ inches, an area which, according to our previous calculations for unidirectional stress, can produce 31 microjoule per deflection. This group has chosen a maximum stress of 20% of the modulus of rupture or $\frac{2}{5}$ of the stress necessary to produce the 31 microjoule output (and a better choice from the fatigue lifetime viewpoint). At this stress about 5 microjoule can be produced per deflection or about the same output as Kennedy has achieved. As did Dr. Kennedy, Dr. Parsonnet has experienced difficulty with fluids leaking through his silicon rubber encapsulation which has limited the electrical lifetime to a few hours.

For an artery clamp of the clothespin type the mechanical work done by each expansion of the artery is roughly proportional to: systolic-diastolic pressure differential, normal variation in arterial diameter in each cycle, arterial diameter, and length of artery used. For the following set of parameters, 20 mm Hg, 2mm, 1cm, 4cm, we calculate an energy yield of 500 microjoule per beat, which is 1 milliwatt at 120 beats/minute. Aortic expansion does appear to be capable of producing the necessary quantity of mechanical work, but it remains for improved mechanical designs to meet the sealing, efficient drive, and adaptation requirements which are demanded of a successful long term system.

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